Echo Time Dependence of Activation - Induced Signal Change Revisited using Diced K-Space (DK) Acquisition

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Introduction:

Experimental results have shown that the TE dependence of BOLD fractional signal change appears linear (1-3) which implies that signal change can be simply described as a change in relaxation rate ($\Delta R2^*$ or $\Delta R2$). Haacke et al. (4) and Yablonski et al. (5) predict that for large vessels, TE dependence of activation-induced fractional signal change may show oscillations as coherent phase shifts destructively and constructively add. Yablonski et al. (5) also develop a model which predicts a non-linear fractional signal change TE dependence.

In this study, the TE dependence of resting and active MR signal is re-evaluated within a wide TE range and with small TE increments using a imaging strategy we call Diced K-space (DK). Following each excitation, one phase encoding gradient amplitude is repeatedly applied for 64 lines of k-space during which the readout gradient rapidly oscillated. This process is repeated 64 times to create a block of data that is reconstructed into 64 images, matrix size = 64 x 64, each image having all k-space lines acquired at the same time during the FID. The first image has a TE of 5.032 ms and the last image has a TE of 44.349 ms. We compare activation - induced signal change TE dependencies with simulations carried out using a deterministic diffusion method(6).

Methods:

DK was performed on one axial imaging plane containing motor cortex using a local three axis gradient coil at 1.5 T (GE Signa). Two subjects were imaged. Voxel volume = 3.8 x 3.8 x 10 mm^3. For DK, TR = 250 ms, flip angle = 45°, TE range = 5.032 ms to 44.349 ms. TE increment for each image = 0.624 ms. TE points (or images per TE curve) = 64. Image acquisition time per decay curve = 16 sec. A total of 96 decay curves were collected: 48 during rest, and 48 during activation. These image sets were registered and averaged.

In the same session, a T2*- weighted arterial spin labeling time series was collected (7-9) to simultaneously map perfusion and BOLD changes. Using these maps as a reference, regions of interest were chosen for evaluation of the DK data.

Results:

Figure 1 shows the MR signal from a region of interest in motor cortex that demonstrated activation - induced BOLD signal changes. A single exponential fit to the curves show T2* values typical for gray matter at 1.5T, and a typical activation - induced $\Delta R2^*$ value of -0.42 s^{-1}. Figure 2.a. shows a non-linear fractional signal change TE dependence. This effect was clearly seen in most regions that demonstrated BOLD signal changes. Only in a region that showed more subtle BOLD changes in combination with robust perfusion change did the fractional signal change appear linear. This bottom curve also showed a zero intercept of 0, while the top curve showed a positive zero intercept - suggestive of inflow effects. This result suggest that perfusion effects seen using arterial spin labeling are not the same as "inflow effects" as seen by a positive zero intercept.

Simulations (6) of the fractional TE dependence were carried out in which either proton diffusion effects were considered insignificant (large vein effects) or significant (capillary effects). Parameters were: resting and active oxygen saturation = 0.6 and 0.75, Bo = 1.5T, $\Delta\chi$ = -0.8 x 10^{-7}, Hct = 40, bv = 5%, radius (R), = 2.5 μm. A random distribution of vessel orientations was used. Diffusion coefficient (D) was set to 0 (large vein effects) and 1 μm2/ms (capillary effects). Figure 2.b shows that, when diffusion effects are not considered, the curve starts out flat then increases slope. When spin diffusion effects are taken into consideration, the curve still starts out more flat and increases slope, but becomes linear relatively rapidly.

Conclusions:

In summary, the simulations show that, as (R^2/D) $\Delta\omega$ decreases, ($\Delta\omega$ = frequency offset of the susceptibility perturbation), the TE dependence becomes increasingly linear at earlier TE values. The implications are that the non-linear TE dependence occurs more with large susceptibility perturbations and more linear TE dependencies arise from small susceptibility perturbations.

DK allows for highly sampled TE decay curves to be rapidly obtained. From these curves, information about vascular geometry contribution to BOLD contrast may be determined on a voxel-wise basis.

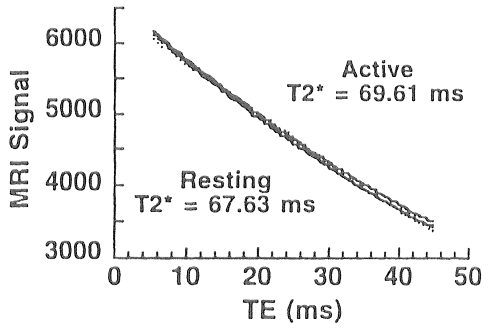
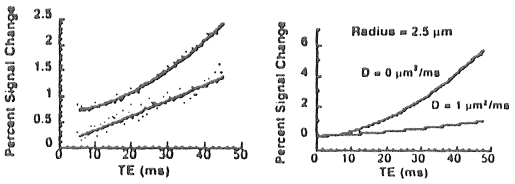


Figure 1: Resting and active signal obtained with DK from a region of interest in motor cortex showing strong BOLD changes.



a. b.

Figure 2: a. Fractional signal change. The top curve was obtained from the decay curves in Figure 1. The bottom curve is from a region that had subtle BOLD changes and clear perfusion changes. **b.** Numerical simulations of fractional signal change TE dependence when diffusion effects are insignificant (large compartments) and when diffusion effects are significant (small compartments). The bottom curve becomes linear more rapidly.

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